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14. ABSTRACT
The purpose of this study was to develop and validate a model with optimal set-up of reflective markers, producing minimal errors in inverse dynamics calculations. The Statement of Work for this project indicated two specific aims. Specific Aim 1 proposed to develop and validate a model with unique optimal marker placements for specific running prosthesis designs. The proposed timeline indicated that preparation for the experimental setting and formulation of the program for data analysis would occur during Months 1-8. These milestones were reached on schedule. During Months 8-16, we proposed to complete MTS testing, begin validating the general model, and begin analyzing the MTS data to determine the final marker model for each running-specific prosthesis. Some of these milestones were delayed due to procurement issues arising from the prosthesis manufacturers. A no-cost extension until 28 February 2012 was granted due to these issues. These milestones were completed as anticipated upon prosthesis procurement. The results of this study indicate that marker placement and number of markers on a running-specific prosthesis did not greatly influence the accuracy of kinetic data of the running prosthesis designs tested. A draft of a manuscript detailing the study results is included as an Appendix to this report.

15. SUBJECT TERMS
Force, Moment, Inverse Dynamics, Biomechanics, Model, Running, Prosthesis

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Introduction

While running has been shown to reduce disease risks and promote a generally healthy lifestyle in uninjured people, very little running-specific research is available pertaining to the amputee population. The little existing amputee running literature primarily involves running with prostheses designed for every day wear, which are typically prescribed and aligned to perform optimally during standing and walking. Further, these studies have used biomechanical models designed for the intact limb to estimate joint kinetics (forces and moments) using an inverse dynamics approach. This approach estimates distal joint kinetics and uses these calculations to estimate more proximal joint kinetics. Consequently, inverse dynamics estimations rely on accurate estimation of the ankle joint as errors will be propagated and inflated with more proximal calculations. The previous studies on amputee running have not validated the methodology used for joint kinetic measurements with running-specific prostheses, which most likely will prove to be erroneous.[1-3] These limitations call for systematic research and development of validated models for running-specific prostheses. This will lead to improved prosthetic designs that will allow clinicians to provide evidence-based exercise prescriptions to amputees, enabling them to comfortably and efficiently run. The objective of the proposed study is to develop and validate a model with optimal set-up of reflective markers used in 3D gait analysis, producing minimal errors in inverse dynamics calculations. The long-term objective of this project is to understand the biomechanical and physiological consequences of amputation, to develop an optimal design of activity-independent lower-extremity prosthesis, and to help clinicians prescribing appropriate prosthesis and exercise regimes to people with a lower extremity amputation.

Please see *Appendix I: Manuscript Draft* for a more detailed introduction.

The current project was approved for funding over 18 months beginning 1 August, 2009 and to be completed by 28 February, 2011. A no-cost extension was approved due to issues in prosthesis procurement caused by the prosthetic manufacturing companies. The extension is approved through 28 February, 2012.

The purpose of this document is to detail progress of the study to satisfy the Annual Report requirement.

Body

The approved Statement of Work proposed the following timeline (Table 1):

Table 1. Timeline for approved project.

	Months 1-6	Months 7-12	Months 13-18
Specific Aim #1: Development and validation of a model with unique optimal marker placements for specific running prosthesis designs			
Formulate program for data analysis	X		
MTS testing of running specific prostheses (12 prostheses)	X		
Validation of model		X	
Analysis of MTS data to determine model		X	
Specific Aim #2: Determination of the resultant optimal marker placement for all tested running prosthetic designs			
Determine model with optimal marker placement for across designs			X

Methodology

A biomechanical model was developed using motion analysis of running-specific prostheses in a material testing system (MTS, Eden Prairie, MN). Four running-specific prosthesis designs (Figure 1) were tested for this project including the 1E90 Sprinter (OttoBock Inc.), Flex-Run (Ossur), Cheetah® (Ossur) and Nitro Running Foot (Freedom Innovations). These prostheses were chosen because they are the most commonly prescribed running-specific prostheses currently available on the market. Three different stiffness categories were also tested for each prosthetic design to identify whether prosthetic stiffness affects optimal marker placement. Stiffness categories were chosen to reflect a common range of stiffnesses that might be prescribed.

Each prosthesis was placed in the MTS between two load cells (Bertec PY6, Columbus, OH) in a neutral alignment. Neutral alignment was defined according to the specific manufacturers' recommendations for prosthesis alignment. The load cells captured data at 1,000 Hz. The prostheses were cyclically loaded for ten cycles with axial forces up to 2,500 N to simulate peak vertical forces commonly observed during running (approximately three times the body weight of a 75 kg person). The load cells measured the force and moment at the point of load application proximal to the prostheses (applied load) and the reaction forces distal to the prostheses (ground reaction forces).

Reflective markers were placed at 2 cm intervals along the lateral aspect of the keel of each running-specific prosthesis (Figure 2). Reflective markers were also placed orthogonally on the anterior, lateral, and medial aspect of the "head" of the prosthesis, at the point of connection to the socket or pylon, in order to define the local coordinate system of the prosthesis. Three additional markers were placed along the midline of each prosthesis to define a plane to which the keel markers were projected for further analysis. An 8-camera motion capture system (Vicon, Oxford, UK) with a capture frequency of 500 Hz was used to collect the 3-D positional data of the markers during each trial. Two consecutive projected center line markers defined individual segments of the prosthesis (assumed to be rigid) and consecutive segments shared a common marker. The joint between these segments was assumed as a hinge joint. Standard inverse dynamics calculations (equations 1 and 2) were used to estimate the force and torque transfer from the ground reaction force, through the defined prosthesis segments, and to the point of load application proximal to the prosthesis.

a. Freedom Innovations Nitro | b. Ossur Flex-Run | c. Ossur Cheetah | d. Ottobock Sprinter 1E90

Figure 1. Prostheses used for mechanical testing.

$$F_i - F_{i+1} + m_i g = m_i a_i \quad [1]$$
$$M_i - M_{i+1} + r_i \times F_i - r_{i+1} \times F_{i+1} = [I_i]\ddot{\theta}_i + \dot{\theta}_i \times ([I_i]\dot{\theta}_i) \quad [2]$$

where F_i and F_{i+1} are forces acting on link *i* at joints *i* (distal) and *i+1* (proximal), respectively; M_i, M_{i+1} are moments exerted on link *i* at joints *i* and *i+1*, respectively; r_i and r_{i+1} are radii from the COM of link *i* to the joint centers *i* and *i+1*, respectively; *g* is acceleration due to gravity; $[I_i]$ is the matrix of inertia; and $\dot{\theta}_i$ and $\ddot{\theta}_i$ are vectors of the angular velocity and acceleration for link *i*, respectively.

The difference between force and moment values at the point of load application from the estimated inverse dynamics calculations and the directly measured values from the top load cell was considered model error. Force and moment estimations were made with every combination of remaining markers giving a resultant error value for each combination. Errors were calculated for each loading cycle as root mean squared error (RMSE) and normalized RMSE (NRMSE), respectively.

Please see *Appendix I: Manuscript Draft* for additional details and figures related to the study methodology.

<u>*Research Accomplishments*</u>

<u>Specific Aim #1</u>

MTS testing of running specific prostheses: Completed
The experimental setup was finalized and MTS testing was performed for the existing running-specific prostheses. A representative example of a prosthesis set up in the MTS machine is shown in Figure 2. Reflective markers were placed along the keel of the prosthesis in 1 cm increments and force transducers are present at the base and top of the experimental setup in order to measure forces and moments at the input (top) and “ground” level.

Figure 2. Running-specific prosthesis (Ossur Flex-Run) setup in the MTS machine.

Formulating program for data analysis: Completed

Once raw experimental data were obtained from the MTS testing, we were able to begin formulating the data analysis program in the Matlab programming language and validate the proposed model. Because the project proposed a new model and method to analyze running-specific prostheses, each stage of the program development required validation to ensure proper measurements and calculations. Consequently, this was a lengthy progress involving a large amount of troubleshooting. The data analysis program has been completed. Please see the data generated from this programming in the Reportable Outcomes section of this document.

Validation of model: Completed
The programming and model validation have been completed for use in the analysis of the MTS data to determine the final marker model.

Analysis of MTS data to determine model: Completed
Determining the final marker model for each specific prosthesis design has been completed. Completion of this task was delayed beyond the originally proposed timeline due to procurement issues with the prosthetic companies. These issues were resolved and the task was completed. This task completes the goals set for Specific Aim 1.

Specific Aim #2

Determine model with optimal marker placement for all prosthesis designs: Completed
This task was proposed for Months 13-18 of the project and required the completion of tasks in Specific Aim 1. This task was delayed due to prosthesis procurement issues. However, the task was completed as proposed.

Results

Data are presented for anteroposterior (AP) force, vertical force, and flexion moment values throughout the cyclical loading task. Figure 3 compares the directly measured values with those estimated from thousands of different combinations of marker placements on the prosthesis. Figure 4 displays the raw error between the estimated force and moment values and the directly measured values for each of the marker placement combinations. Figure 5 shows the NRMSE, representing the difference (in percent) between the directly measured (via load cell) and estimated (via inverse dynamics) proximal forces and moments calculated for each marker combinations.

The Freedom Innovations Nitro prosthesis had a maximal RMSE range of 0.26 N (AP force), 4.45 N (vertical force), and 1.02 Nm (flexion moment) and a maximal NRMSE range of 0.02%, 0.17%, and 0.86% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ossur Flex-Run prosthesis had a maximal RMSE range of 4.37 N (AP force), 5.88 N (vertical force), and 1.05 Nm (flexion moment) and a maximal NRMSE range of 0.37%, 0.28%, and 0.56% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ossur Cheetah prosthesis had a maximal RMSE range of 0.99 N (AP force), 9.38 N (vertical force), and 0.73 Nm (flexion moment) and a maximal NRMSE range of 0.12%, 0.44%, and 0.53% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ottobock 1E90 prosthesis had a maximal RMSE range of 0.48 N (AP force), 7.54 N (vertical force), and 0.54 Nm (flexion moment) and a maximal NRMSE range of 0.07%, 0.35%, and 0.31% for AP force, vertical force,

and flexion moment, respectively across all stiffness categories and all tested combinations of markers.

Figures 6-8 show the average RMSE values for AP force, vertical force, and flexion moment, respectively, for each prosthesis across the number of markers on the prosthesis. All tested combinations with the number of markers indicated were averaged to generate each data point.

Cumulatively, these data indicate little difference in kinetic calculations between the directly measured values and any marker placement or combination of markers.

Please see *Appendix I: Manuscript Draft* for a more detailed report of the study results.

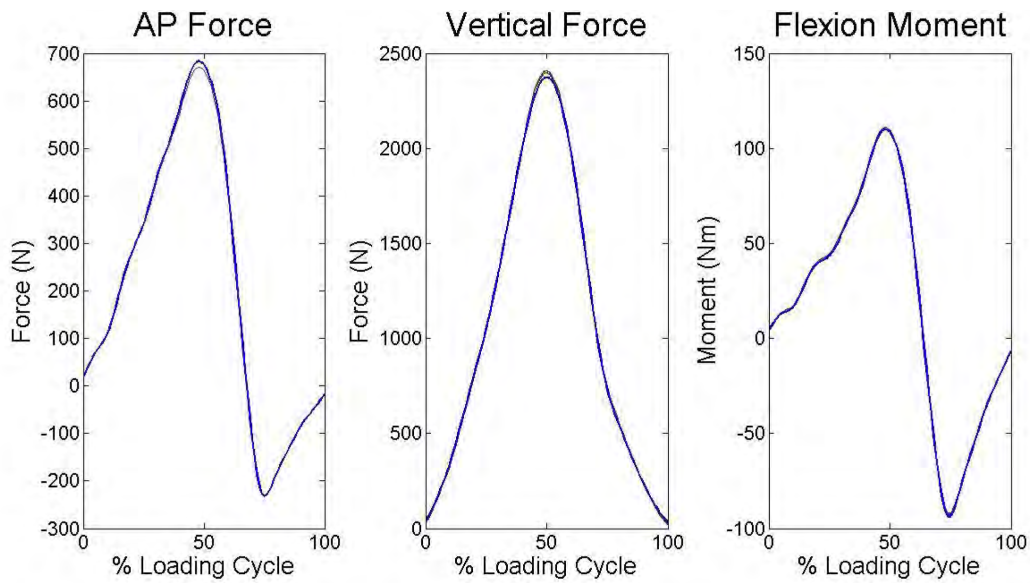

Figure 3. AP force, vertical force, and flexion moment curves for cyclical loading. Thick blue lines represent the directly measured values from the upper load cell. Thin lines represent calculated values from each different combination of markers. Exemplar data is from the Ossur Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

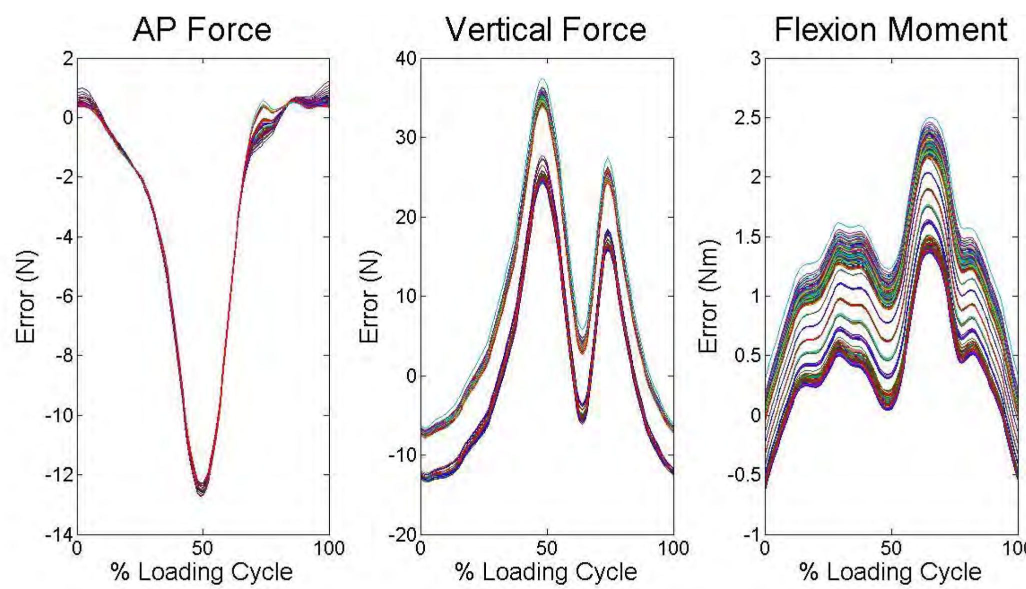

Figure 4. Average AP force, vertical force, and flexion moment error curves for the loading cycle. Each curve represents the difference between the directly measured values from the upper load cell and calculated values from each combination of markers on the prosthesis. Exemplar data is from the Ossur Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

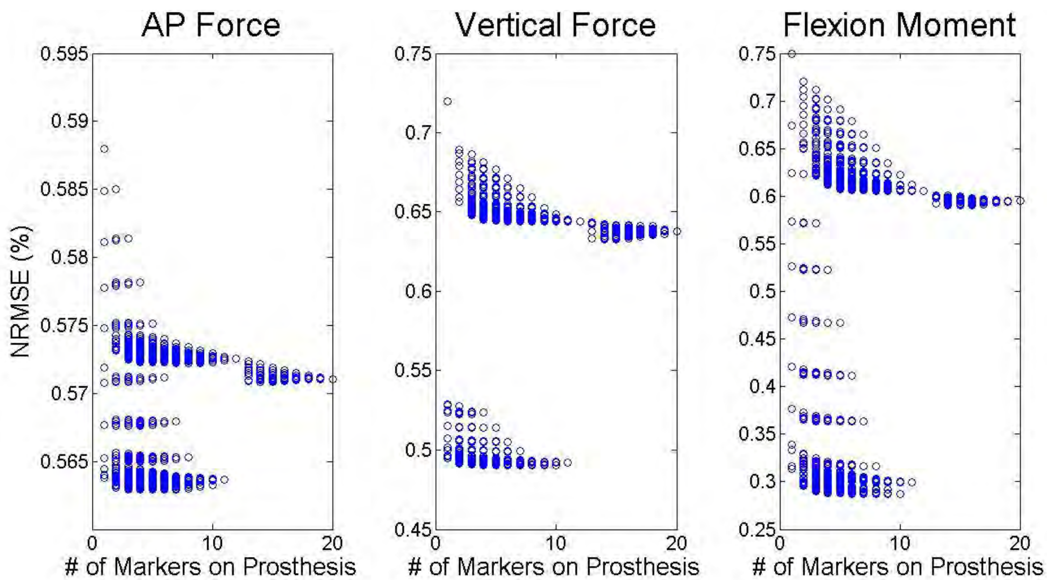

Figure 5. Normalized root mean square error (NRMSE) for each combination of markers for AP force, vertical force, and flexion moments throughout the loading cycle. Each dot represents the NRMSE value for a particular combination of markers. The x-axis shows the number of markers on the prosthesis for the particular combination. Exemplar data is from the Ossur Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

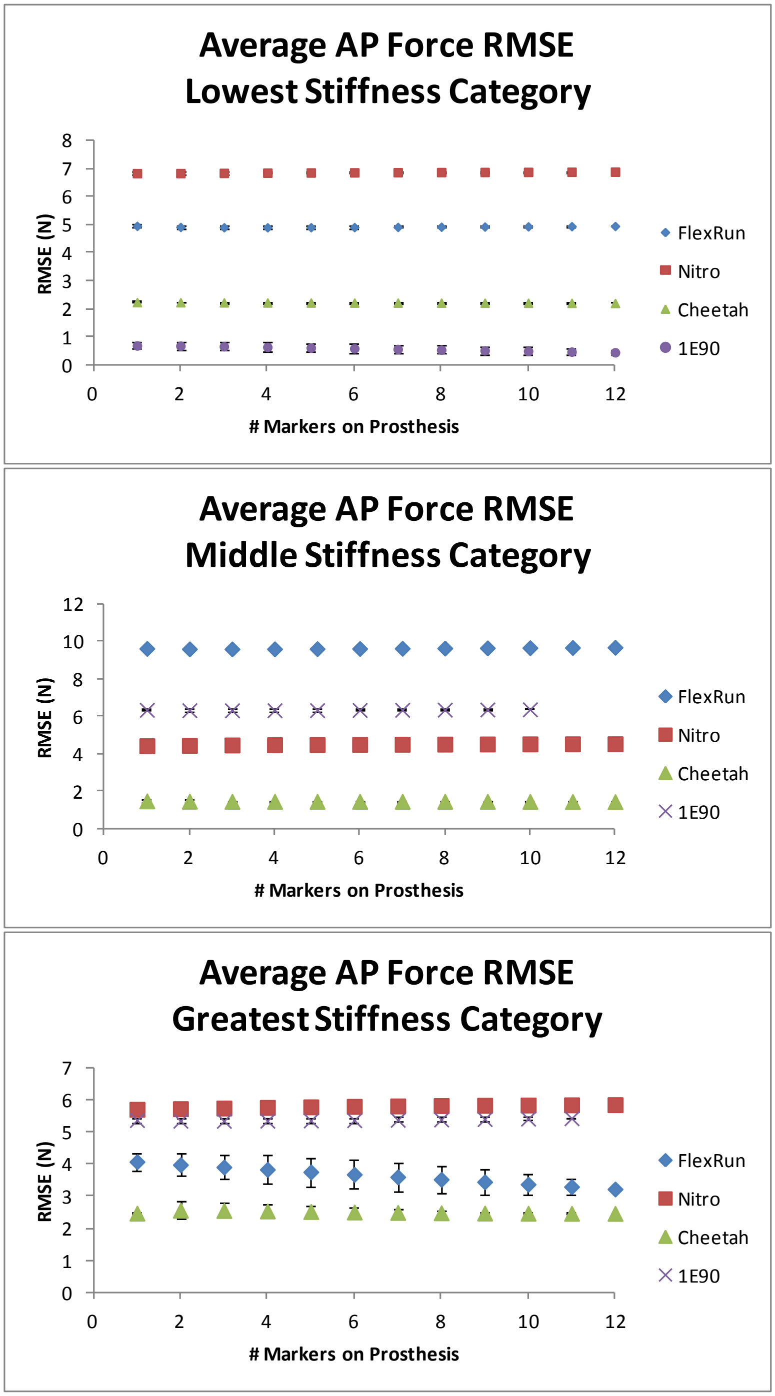

Figure 6. Average anteroposterior (AP) force root mean squared error (RMSE) for each prosthesis across the number of markers on the prosthesis. All tested combinations with the number of markers indicated were averaged to generate each data point. Error bars represent ±1 standard deviation of all marker combinations tested for the number of markers shown.

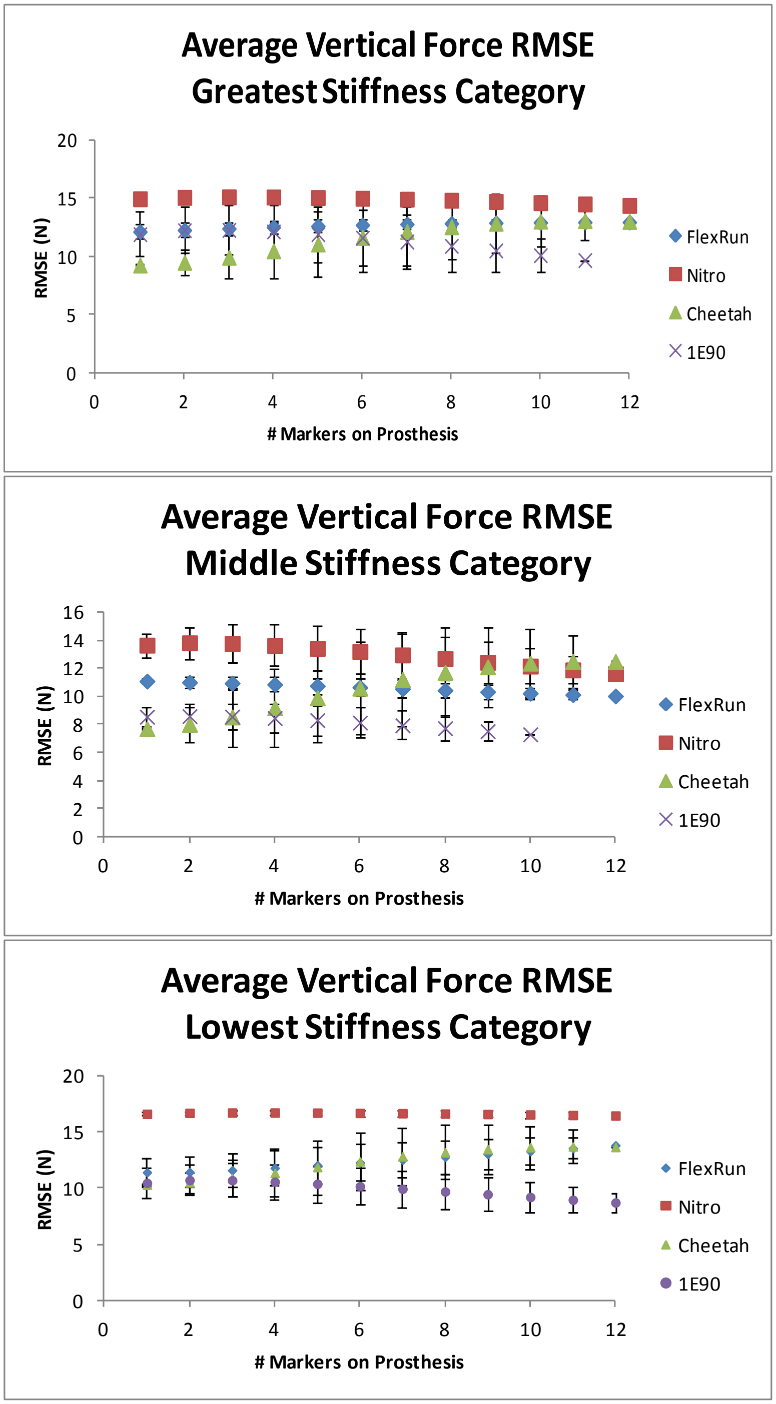

Figure 7. Average vertical force root mean squared error (RMSE) for each prosthesis across the number of markers on the prosthesis. All tested combinations with the number of markers indicated were averaged to generate each data point. Error bars represent ±1 standard deviation of all marker combinations tested for the number of markers shown.

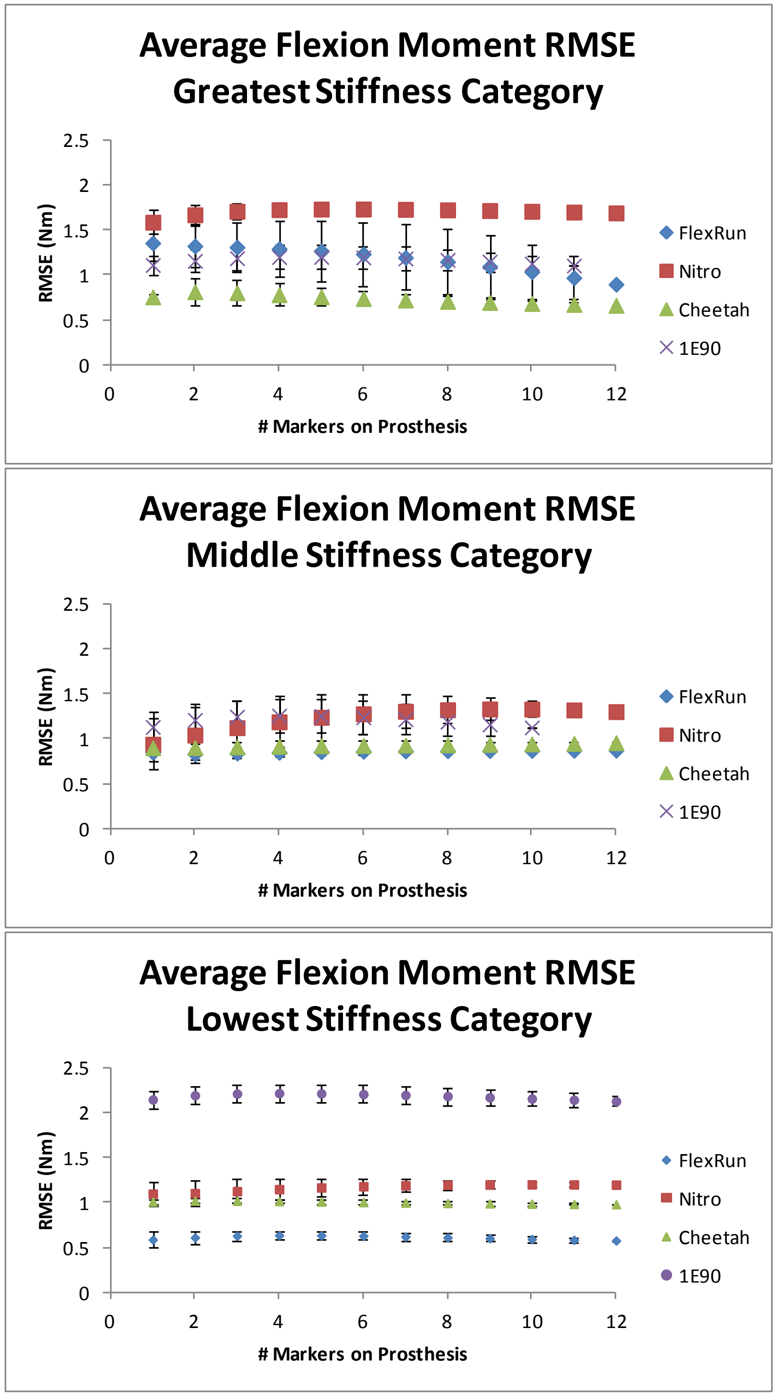

Figure 8. Average flexion moment root mean squared error (RMSE) for each prosthesis across the number of markers on the prosthesis. All tested combinations with the number of markers indicated were averaged to generate each data point. Error bars represent ±1 standard deviation of all marker combinations tested for the number of markers shown.

Key Research Accomplishments

- Experimental setup and testing protocol were completed.
- Formulation of the program for data analysis was completed.
- Validation of the program and model was completed.
- Determining the final marker model for each specific prosthesis design was completed.
- Determining a resultant marker model for all tested running prosthesis designs was completed.
- This research determined that neither the number of markers placed nor the placement of the markers on a running-specific prosthesis influence the force and torque transfer estimations to the proximal end of the prosthesis.
- A manuscript for journal publication is being finalized for submission. A draft of this manuscript is included as Appendix I of this document.

Reportable Outcomes

Manuscripts Supported by this Award

1. Baum BS, Koh K, Linberg A, Tian A, Kim H, Hsieh A, Wolf EJ, Shim JK. (In Preparation) *Optimization and validation of a biomechanical model for analyzing running-specific prostheses*.

Meeting Abstracts Supported by this Award

1. Baum BS, Shim JK. (2009) *Optimization and validation of a biomechanical model for running-specific prostheses*. Research Interaction Day, University of Maryland, College Park (September 18, College Park, MD).
2. Baum BS, Borjian R, Kim YS, Linberg A, Shim JK. (2010) *Optimization and validation of a biomechanical model for analyzing running-specific prostheses*. The 26^{th} Southern Biomedical Engineering Conference (April 30-May 2, College Park, MD).
3. Baum BS, Borjian R, Linberg A, Koh K, Shim JK. (2011) *Optimization and validation of a biomechanical model for running-specific prostheses*. The 15th Annual Meeting of the Gait and Clinical Movement Analysis Society (April 26-29, Bethesda, MD).

Funding Applied for Based on Work Supported by this Award

1. Shim JK (PI), Baum BS (AI, Project Coordinator), *A New Biomechanical Model to Examine Joint Control Adaptations during Running in Individuals with Lower Extremity Amputation*, National Institute of Arthritis and Musculoskeletal and Skin Disease (NIAMS) R03 Award, $141,390.

Conclusions

The research project has been completed as proposed and a manuscript with the final data is in preparation. Some unexpected difficulties (e.g. procurement of prostheses; detailed in the August 2010 Annual Report for this project) delayed portions of the research in the first year of the project. A no-cost extension was granted based on these issues. The issues were resolved and did not affect the overall successful completion of the project.

The data shown in the Body section allowed us to identify the iterations (combinations of marker placements) that yielded acceptable error between the estimated (via inverse dynamics calculations) and directly measured (via load cell) force and moment values. These data indicate that marker placements on running-specific prostheses can be flexible if only kinetic analyses are desired; however, more specific marker placements are recommended if kinematic information about the prosthesis compression is of interest. These data will guide future research and provide greater confidence in reported kinetic results in the past literature.

The data indicate that the marker combinations tested result in errors of less than 1.9% for force and moment calculations for all running prosthesis designs. These data suggest that placing one marker on a running-specific prosthesis is sufficient for accurate joint kinetic analyses. This knowledge allows a larger number of research laboratories to perform running analyses since fewer motion capture cameras would be needed to perform the analysis. This will also dramatically reduce the setup time (fewer markers = less time spent during setup) and the impact on the individual being tested.

Please see *Appendix I: Manuscript Draft* for a more detailed report of the study conclusions.

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Appendix I. Manuscript Draft

Optimization and Validation of a Biomechanical Model for Analyzing Running-Specific Prostheses

Abstract

Modeling the ankle joint during amputee locomotion provides a great challenge since a definitive joint axis may not exist within the prosthetic foot design. Gait analysis estimates joint center positions and defines body segment motions by placing reflective markers on anatomical landmarks. Inverse dynamics techniques then estimate joint kinetics (forces and moments) and mechanical energy expenditure using data from ground reaction forces (GRFs) and the most distal joint (usually the ankle) to make calculations for proximal joints. Running-specific prostheses (RSPs) resemble a "C" or "L" shape rather than the human foot. This allows RSPs to flex and return more propulsive energy, like a spring, but no "ankle" exists. Current biomechanical models assume such a joint exists by placing markers arbitrarily on the RSP (e.g. the most acute point on the prosthesis curvature). These models are not validated and may produce large errors since inverse dynamics assumes rigid segments between markers but RSPs are designed to flex. Moreover, small errors in distal joint kinetics calculations will propagate up the chain and inflate errors at proximal joints.

This study developed and validated a model for gait analysis with RSPs. Reflective markers were placed 2 cm apart along the lateral aspects of four different RSP models with three different stiffness categories each (12 total RSPs). Prostheses were neutrally aligned in a material testing system between two load cells. Forces simulating peak running loads were applied and the load cells measured forces and moments at the proximal (applied force) and distal (GRF) ends of the prostheses. Inverse dynamics estimated force transfers from the ground

to the proximal endpoint of the prostheses through the segments defined by reflective markers. Differences between estimated and applied values at the proximal endpoint were considered model error. Error was calculated for every combination of markers to determine the minimal marker set with an "acceptable" level of error. The results indicate that placing a single marker on an RSP is sufficient for accurate stance phase kinetic analyses.

Introduction

Modeling the lower extremity joints, and specifically the ankle joint, proves to be continual source of difficulty and remains as an inherent problem analyzing locomotion (walking and running) of individuals with lower extremity amputations (ILEA). Identifying the ankle joint during biomechanical analyses of human locomotion is one of the most important tasks because calculations of joint kinetics (forces and moments) and joint mechanical energy start from the ankle joint. A small joint position error at the ankle can easily propagate to the knee and hip joints producing greater errors more proximally. Many of today's commonly prescribed prosthetic foot designs are either energy storage and return (ESAR) or dynamic response feet, which are designed for walking and have a resemblance to an intact foot. During a three-dimensional gait analysis, reflective markers are placed on anatomical landmarks to estimate the positions of joint centers and to define the body segment motions. However, in locomotion studies using walking-specific prostheses, markers defining the ankle joint axis are often affixed to spots on the prosthetic foot that mimic the relative marker location on the intact foot and ankle complex[4-9]. Researchers will often treat current prostheses like an intact limb even though these devices may not have the same architecture or landmarks.

With the development of running specific prostheses, new prosthetic foot designs have emerged that no longer resemble the human foot. Many of the designs resemble a “C” or “L” shape at the distal end of the limb, which allows the prosthesis to flex and return more energy for propulsion during running, similar to a spring. These designs do not have a typical ankle joint (Figure 1); however, many researchers analyze these prostheses using similar methods of biomechanical analyses as have been employed in ESAR and dynamic response prosthetic feet, traditional prosthetic feet (such as SACH and single-axis feet), and the intact limb. Studies investigating running with these devices have estimated the prosthetic limb ankle joint to be either at the same relative position as the intact limb’s ankle joint or the most acute point on the prosthesis curvature (i.e., the greatest curvature; see Figure 1)[9-11]. These estimations have not been validated and potentially result in large errors within the kinetic calculations and subsequent interpretations of results. Using the intact limb as a reference for marker placement also excludes such a model from use on ILEA with bilateral amputations. Consequently, improved and validated modeling techniques are needed to estimate accurate centers of rotation for running prostheses that can be applied to multiple prosthetic designs, and can be utilized in

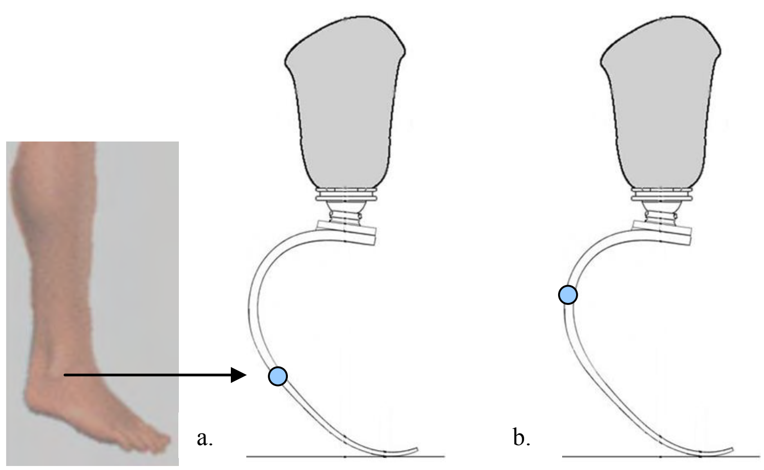

Figure 1. Literature has reported marker placement for running prostheses (a) placed at the height of the intact limb’s lateral malleolus or (b) the point at which the radius of the prosthesis is most acute.

those with bilateral lower extremity amputations where an intact ankle joint is not available for reference. An accurate model will provide data that can be interpreted with confidence and is needed to produce biomechanical and physiological data necessary to identify optimal running techniques, prosthetic alignment, prosthetic designs, training regimens, and energy efficiency. Understanding the biomechanical and physiological consequences of exercise after amputation will allow clinicians to prescribe more appropriate prostheses and exercise regimes to people with a lower extremity amputation.

In addition to the different designs of running-specific prostheses, each of these devices are manufactured in different stiffness categories that are generally prescribed based on an individual's body weight. A heavier person is typically prescribed a RSP with a higher category of stiffness (higher categories correspond to greater prosthesis stiffness). Studies investigating prosthesis stiffness indicate that the stiffness affects performance and body weight alone may be insufficient for prescribing a stiffness category. A stiffer forefoot, wider c-curve, and thinner lay-up resulted in ILEA running their fastest sprint times[12], which suggests that sprint speed can be a function of stiffness and prosthetic foot shape[13]. Using a greater category of stiffness may also improve gait symmetry values for ILEA with transtibial amputation[14], but it has also been shown to reduce energy efficiency[15]. These data suggest that different prosthesis stiffness categories could affect the performance of the prosthesis and therefore the force and torque transfer through the device.

The aim of this experiment was to develop and validate a model with unique optimal marker placements for specific running prosthesis designs and stiffness categories in order to improve the accuracy of future research on RSPs. We hypothesized that more markers placed on any running-specific prosthesis with any stiffness category would result in the least error in

proximal force and moment estimations. Our second hypothesis was that shape would primarily affect the force and torque transfer, so different prosthetic designs would have different optimal marker placements for kinetic analyses, but different stiffness categories within a specific prosthetic design would not affect the optimal marker placement.

Methods

A biomechanical model was developed using motion analysis of running-specific prostheses in a material testing system (MTS, Eden Prairie, MN). Four running-specific prosthesis designs were tested for this project including the 1E90 Sprinter (OttoBock Inc.), Flex-Run (Ossur), Cheetah® (Ossur) and Nitro Running Foot (Freedom Innovations) (Figure 2). These prostheses were chosen because they are the most commonly prescribed running-specific prostheses currently available on the market. Three different stiffness categories were also tested for each prosthetic design to identify whether prosthetic stiffness affects optimal marker placement. Stiffness categories were chosen to reflect a common range of stiffnesses that might be prescribed. For the Flex-Run and Cheetah models, stiffness categories 3, 5, and 7 were tested.

a. Freedom Innovations Nitro   b. Ossur Flex-Run   c. Ossur Cheetah®   d. Ottobock Sprinter 1E90

Figure 2. Prostheses used for mechanical testing.

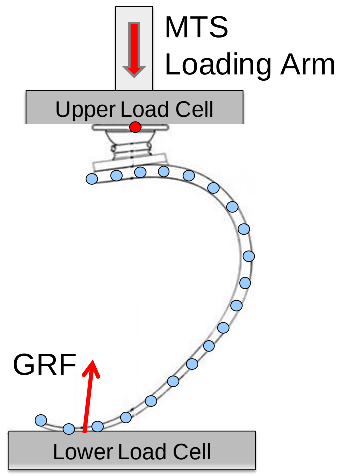

Figure 3. Marker placement on a running-specific prosthesis and its position in an MTS machine between two load cells. Fewer markers than actual are shown in the illustration for clarity. The red dot indicates the point of load application, measured by the upper load cell. The lower load cell measured ground reaction force (GRF). The red arrows represent the input and GRF force vectors.

For the Nitro model, stiffness categories 3, 6, and 7 were tested, and for the 1E90 model, prostheses designed for individuals of 140 lb (63.6 kg), 185 lb (84.1 kg), and 235 lb (106.8 kg) were tested. OttoBock does not use the term "category" to reflect stiffness, rather different prosthesis stiffnesses are reflected by the target weight of the person using the device. Each prosthesis was placed in the MTS between two load cells (Bertec PY6, Columbus, OH) in a neutral alignment (Figure 3). Neutral alignment was defined according to the specific manufacturers' recommendations for prosthesis alignment. The load cells captured data at 1,000 Hz. The prostheses were cyclically loaded for ten cycles with axial forces up to 2,500 N to simulate peak vertical forces commonly observed during running[16-18] (approximately three times the body weight of a 75 kg person). The load cells measured the force and moment at the point of load application proximal to the prostheses (applied load) and the reaction forces distal to the prostheses (ground reaction forces).

Reflective markers were placed at 2 cm intervals along the lateral aspect of the keel of each running-specific prosthesis (see Figure 3). Reflective markers were also placed

orthogonally on the anterior, lateral, and medial aspect of the "head" of the prosthesis, at the point of connection to the socket or pylon, in order to define the local coordinate system of the prosthesis. Three additional markers were placed along the midline of each prosthesis to define a plane to which the keel markers were projected for further analysis. An 8-camera motion capture system (Vicon, Oxford, UK) with a capture frequency of 500 Hz was used to collect the 3-D positional data of the markers during each trial. Two consecutive projected center line markers defined individual segments of the prosthesis (assumed to be rigid) and consecutive segments shared a common marker. The joint between these segments was assumed as a hinge joint. Standard inverse dynamics calculations[19] were made to estimate the force and torque transfer from the ground reaction force, through the defined prosthesis segments, and to the point of load application proximal to the prosthesis.

Prosthesis thickness and width were measured at each marker position using digital calipers. Prosthesis segments were defined by two consecutive markers and were considered as rigid trapezoidal cuboids (see Figure 4). The center of mass along the width and thickness of each segment were determined from half the average width and thickness, respectively. The center of mass position along the long axis (length) of each segment was determined by equation 3:

$$\frac{w_d+2w_p}{3(w_p+w_d)} * l \qquad [3]$$

where w_d and w_p are the distal and proximal end widths and l is the segment length.

The inertial properties of each prosthesis segment were estimated using assumptions based on a trapezoidal cuboid. Each segment length was integrated across 200 subsegments. The principal axis moments of inertia of each segment were estimated by equations 4-6:

$$I_{xx} = \sum_1^i [\tfrac{1}{12} m_i (l_i^2 + w_i^2) + m_i r_i^2] \qquad [4]$$

$$I_{yy} = \sum_1^i [\tfrac{1}{12} m_i (l_i^2 + t_i^2) + m_i r_i^2] \qquad [5]$$

$$I_{zz} = \sum_1^i [\tfrac{1}{12} m_i (w_i^2 + t_i^2) + m_i r_i^2] \qquad [6]$$

where m_i is the mass, l_i is the length, w_i is the width, t_i is the thickness, r_i is the distance between the subsegment center of mass and the segment center of mass for each subsegment i, respectively.

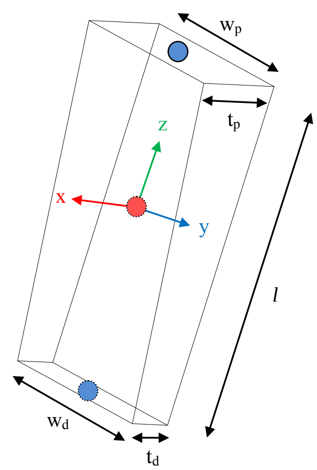

Figure 4. Schematic of segment definitions within each prosthesis. Blue circles represent markers, the red circle represents the segment center of mass. The axis defines the segment local coordinate system with its origin at the center of mass. Segment length (l) is also shown along with the width (w) and thickness (t) at the proximal (p) and distal (d) ends.

The angles between each set of three consecutive markers were calculated throughout the cyclic loading cycles and the range of angle change was determined at each marker "joint". Markers that did not have an angular change of greater than one degree were removed from further analyses as they were considered as part of a larger rigid segment. The remaining markers were used for the model analysis. This process was performed to reduce the total number of marker placement combinations for the analysis. For example, if 25 markers were placed on a prosthesis, the total number of possible combinations to examine would equal 2^{25} (over 33.5 million). After applying the angular change threshold of one degree, if the remaining markers reduced to 12, this would provide 2^{12}, or 4096, possible combinations of marker placement to test.

The difference between force and moment values at the point of load application from the estimated inverse dynamics calculations and the directly measured values from the top load cell was considered model error. Force and moment estimations were made with every combination of remaining markers giving a resultant error value for each combination. Error was calculated for each loading cycle using Equations 7 and 8 for root mean squared error (RMSE) and normalized RMSE (NRMSE), respectively.

$$RMSE = \frac{\sum(K_m - K_e)^2}{n} \qquad [7]$$

$$NRMSE = \frac{RMSE}{(max_{Km} - min_{Km})}\% \qquad [8]$$

where K_m represents the directly measured kinetic values from the upper load cell, K_e represents the estimated kinetic values from inverse dynamics equations, n is the number of data points in the loading cycle, and *max* and *min* represent the maximum and minimum values within the loading cycle, respectively.

These error values were analyzed to determine an “acceptable” level of error for a minimal marker set that can be used by most motion capture laboratories. Less than 5% error from the peak force and moment values was considered acceptable.

Results

Calculated values and error data are presented for anteroposterior (AP) forces, vertical forces, and flexion moments during the cyclical loading trials for each prosthesis. Mediolateral (ML) forces, ML rotational moments, and internal/external rotational moments are not presented since the axial loading of the prostheses produced minimal forces and moments along and about these axes, respectively.

Regardless of the number of markers or their placement on the various RSPs, force and moment calculations using inverse dynamics techniques resulted in errors of less than 1.6% as compared to the directly measured values (Table 1). Directly measured and calculated AP force, vertical force, and flexion moment values are presented in Figure 5. Raw errors and NRMSE between the directly measured and calculated forces and moments are presented in Figures 6-7.

The Freedom Innovations Nitro prosthesis had a maximal RMSE range of 0.26 N (AP force), 4.45 N (vertical force), and 1.02 Nm (flexion moment) and a maximal NRMSE range of 0.02%, 0.17%, and 0.86% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ossur Flex-Run prosthesis had a maximal RMSE range of 4.37 N (AP force), 5.88 N (vertical force), and 1.05 Nm (flexion moment) and a maximal NRMSE range of 0.37%, 0.28%, and 0.56% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ossur Cheetah prosthesis had a maximal RMSE range of 0.99 N (AP force), 9.38 N

(vertical force), and 0.73 Nm (flexion moment) and a maximal NRMSE range of 0.12%, 0.44%, and 0.53% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers. The Ottobock 1E90 prosthesis had a maximal RMSE range of 0.48 N (AP force), 7.54 N (vertical force), and 0.54 Nm (flexion moment) and a maximal NRMSE range of 0.07%, 0.35%, and 0.31% for AP force, vertical force, and flexion moment, respectively across all stiffness categories and all tested combinations of markers.

Table 1. Error ranges (minumum to maximum RMSE and NRMSE) of all combinations of markers for the estimated kinetic values from inverse dynamics equations.

		Freedom Innovations Nitro			Ossur Flex-Run			Ossur Cheetah			Ottobock 1E90		
Stiffness Category:		Cat 3	Cat 6	Cat 7	Cat 3	Cat 5	Cat 7	Cat 3	Cat 5	Cat 7	140 lb	185 lb	235 lb
AP Force	RMSE (N)	6.78-6.92	4.32-4.50	5.66-5.92	5.17-9.54	9.39-9.54	3.23-4.21	2.17-2.27	1.36-1.59	2.46-3.45	0.29-0.77	6.27-6.46	5.28-5.56
	NRMSE (%)	0.68-0.69	0.61-0.63	0.54-0.56	0.56-0.93	0.92-0.93	0.35-0.45	0.31-0.32	0.34-0.40	0.29-0.41	0.05-0.12	1.43-1.47	0.90-0.95
Vertical Force	RMSE (N)	16.37-16.80	11.55-16.00	14.39-15.85	11.05-16.93	10.19-11.95	11.85-14.18	10.27-17.41	7.57-16.95	9.22-17.49	6.88-11.93	7.17-9.61	7.16-14.70
	NRMSE (%)	0.64-0.66	0.45-0.62	0.50-0.55	0.44-0.72	0.41-0.48	0.49-0.59	0.30-0.50	0.36-0.80	0.28-0.53	0.41-0.71	0.47-0.63	0.33-0.68
Flexion Moment	RMSE (Nm)	0.98-1.36	0.81-1.83	1.27-1.99	0.63-1.59	0.78-1.09	0.89-1.94	0.91-1.13	0.87-1.14	0.66-1.39	2.02-2.32	0.98-1.52	1.03-1.38
	NRMSE (%)	0.52-0.71	0.67-1.53	0.72-1.14	0.31-0.78	0.37-0.51	0.48-1.04	0.67-0.83	0.53-0.70	0.46-0.99	0.75-0.86	0.58-0.89	0.38-0.51

*Notes: RMSE = root mean square error, NRMSE = normalized root mean square error

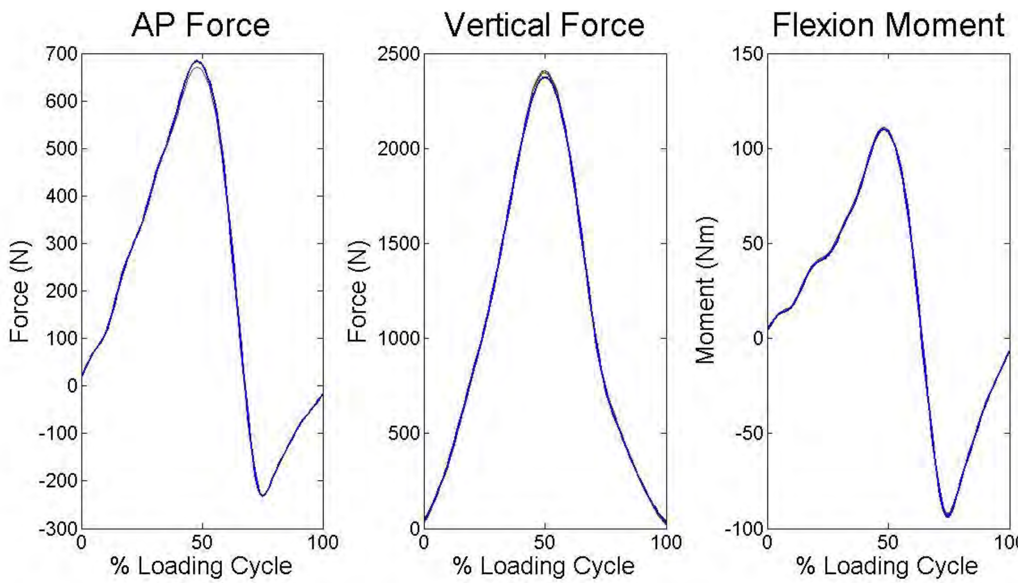

Figure 5. Anteroposterior (AP) force, vertical force, and flexion moment curves for cyclical loading. Thick blue lines represent the directly measured values from the upper load cell. Thin lines represent calculated values from each different combination of markers. Exemplar data is from the Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

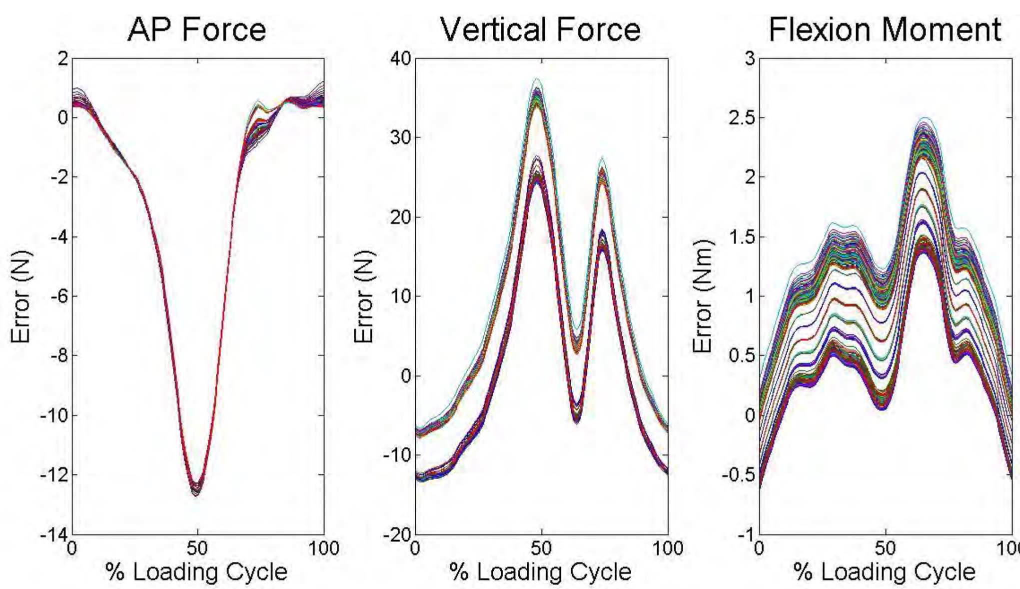

Figure 6. Average anteroposterior (AP) force, vertical force, and flexion moment error curves for the loading cycle. Each curve represents the difference between the directly measured values from the upper load cell and calculated values from each combination of markers on the prosthesis. Exemplar data is from the Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

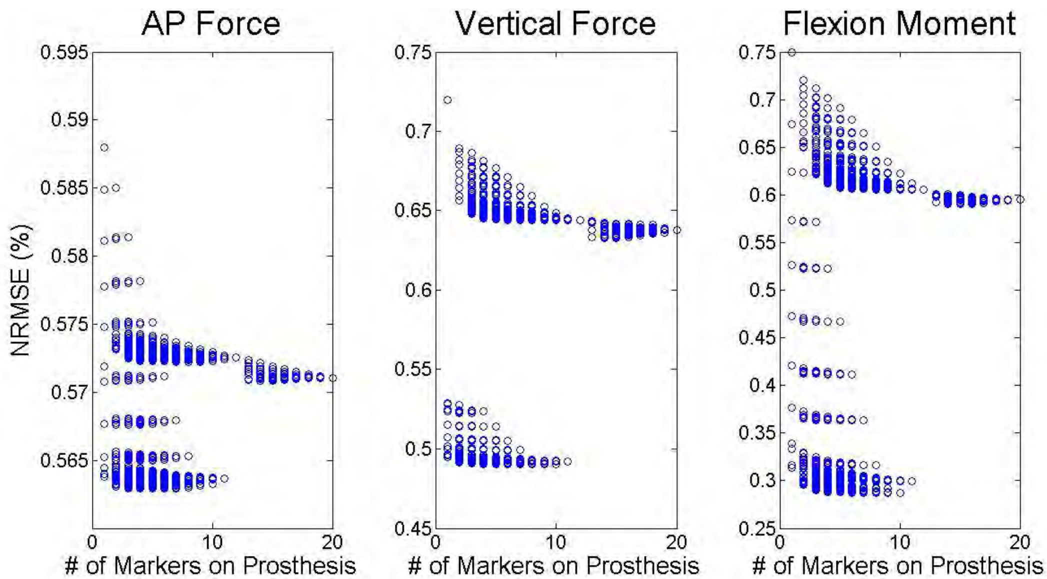

Figure 7. Normalized root mean square error (NRMSE, %) for each combination of markers for AP force, vertical force, and flexion moments throughout the loading cycle. Each dot represents the NRMSE value for a particular combination of markers. The x-axis shows the number of markers on the prosthesis for the particular combination. Exemplar data is from the Flex Run category 3 prosthesis. Other tested prostheses and stiffness categories showed similar results.

Discussion

The data from this study suggest that the number and placement of markers on any of the tested running-specific prostheses does not greatly influence the estimation of force and moment transfer through the prostheses. The magnitude of the ground reaction forces is very large in comparison to the inertial properties of the running-specific prostheses. Therefore, these ground reaction forces generate torques that account for nearly all of the estimated moments while the moments of inertia and the angular velocities of the prosthetic segments contribute relatively little to stance phase kinetics.

These data suggest that kinetic data calculated from prior research with RSPs may be interpreted with greater confidence. Placing markers at the same relative position as the intact

limb's ankle joint or the most acute point on the prosthesis curvature[9-11] will yield similar results in resultant kinetic values proximal to the prosthesis. However, for consistency and flexibility in modeling, it is recommended that markers are placed according to the prosthesis architecture rather than intact limb architecture. This will allow markers to be placed on the same location of a particular prosthesis from subject to subject and will allow for the study of ILEA with bilateral amputations.

As a minimal marker set, we recommend placing markers at the most proximal/frontal position on the prosthesis (on the "head" of the RSP), at the most acute point on the prosthesis curvature, and at the most distal/frontal position on the prosthesis (on the "toe" of the RSP). These markers outline the dimensions of the prosthesis. Placing additional markers on the prostheses will allow for a more detailed description of prosthetic kinematics; however, they have little effect on stance phase kinetic estimations.

Most motion capture laboratories have a limited number of cameras and may have difficulty tracking a large number of markers placed closely together during activities such as running. This limits the number of markers that researchers can feasibly place on the keel of a running-specific prosthesis, especially considering that the thin profile of such prostheses often necessitates using markers with small diameters. Furthermore, motion capture of overground running requires a large capture volume, and optimizing camera placement for large volumes reduces the effectiveness of these systems to capture small markers in close proximity to each other. Utilizing a minimal marker set for running-specific prostheses will enable widespread use of such a model regardless of the number of cameras available to a laboratory and to allow for both overground and treadmill data collections while using the same model. Additionally, fewer markers on a prosthesis makes setup less tedious and saves testing time.

Several limitations exist in this study. First, only axial loading was performed on the prostheses, whereas when running, the prostheses are loaded while rolling forward, which would produce different loading patterns and prosthetic bending. This could affect the recommended marker placements on the prostheses. However, the overall ground reaction forces during running are still much larger than the inertial properties of the prostheses, so it is anticipated that for kinetic analyses, the results presented in this study would generalize to overground running. However, due to the axial loading, this study only presented AP force, vertical force, and flexion moment results. Validation of the marker models is still needed for mediolateral forces, varus/valgus moments, and internal/external rotational moments and would require a 6-degree-of-freedom material testing system that could mimic the prosthetic roll-over during running or direct load measurements at the proximal end of the prosthesis during running. An additional limitation of this study is that only stance phase was investigated. The inertial effects of the running prostheses during swing phase are most likely not trivial, so accurate measures of mass, center of mass position, and moments of inertia are needed to accurately estimate the joint kinetic values proximal to the prostheses. Future studies are needed to accurately measure and predict the inertial properties and effects of running-specific prostheses during the running swing phase.

The development and validation of an accurate biomechanical model for use with running-specific prostheses allows researchers to fully examine the kinematic and kinetic adaptations that occur during running in ILEA. Extremely limited information is available in the literature to guide clinicians in aligning, prescribing, or rehabilitating ILEA who wish to run. For example, it is currently unknown whether running with running-specific prostheses poses an

increased risk for injury in the residual limb joints or joints in the contralateral limb. ILEA are already at greater risk of degenerative joint diseases such as osteoarthritis (OA)[20], and the larger forces generated during running could promote the development and progression of these diseases. Prior research supports that OA may initiate in joints that experience a traumatic or chronic event (such as amputation due to injury or disease) that causes kinematic changes[21]. The rate of OA progression is currently thought to be associated with increased loads during ambulation[21, 22]. Identifying running techniques, prosthetic alignments, or new prosthetic designs that reduce peak lower extremity joint loading may reduce the risk of developing and progressing OA.

Additional research needs include investigating the effects of various prosthetic components in meeting different running goals, and determining optimal prosthetic alignment so as to minimize asymmetries and maximize energy efficiency during running.

Conclusions

Regardless of the number of markers or their placement on the various RSPs, force and moment calculations using inverse dynamics techniques resulted in errors less than 1.9% as compared to the directly measured values at the proximal end of the prostheses. This affords researchers the flexibility to place markers conveniently on running-specific prostheses and still confidently estimate joint kinetic data during the stance phase of running.

A validated biomechanical model is necessary to aid in our analysis and knowledge of the effects of using running-specific prostheses. Development of this model allows researchers to systematically analyze the kinematic and kinetic adaptations of individuals with lower extremity amputations during running. This information will lead to improved prosthetic prescription and

alignment, rehabilitation techniques, and prosthetic designs that will improve performance and reduce risks for injury and disease.

Acknowledgements

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